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published in
IEA 2000 / HFES 2000 Congress
2000

2D ANALYSIS OF 3D LIFTING: HOW FAR CAN WE GO ?

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Occupational manual material handling (MMH) is generally not limited to the sagittal plane. Yet, for practical reasons, biomechanical modeling of low back loading during occupational MMH is mostly restricted to 2D. In this study, the limitations to such an approach are analyzed through quantification of the errors made during 2D analysis of 3D lifting. In addition, an estimate is given of the improvements that can be obtained using a simple method to resolve one major source of error, i.e. the error due to projection of lumbar markers onto the sagittal plane.

INTRODUCTION

It has been accepted that there is a positive association between occupational low back loading and the occurrence of low back pain. In recent epidemiological reviews it is concluded that the strength of this association is highly dependent on the quality of the instrument that is used to measure mechanical loading at the low back (Burdorf & Sorock, 1997; Ferguson & Marras, 1997; NIOSH, 1997). Therefore, direct measurements seem to be preferable over questionnaires or observation techniques. The most widely spread method to quantify low back loading is the application of linked segment models (Chaffin, 1969). However, large scale monitoring of low back loading at the workplace is difficult and time-consuming, even when the linked segment analysis is static and restricted to the sagittal plane (Norman et al., 1998). The magnitude of the underestimation of low back loading through static analysis depends on the speed of lifting and may, expressed as a percentage of measured dynamic torque, exceed 40 % for fast lifting movements (Looze, Kingma, Thunissen, Wijk, & Toussaint, 1994). In smaller studies the use of dynamic models may solve this problem (Looze et al., 1995; Looze et al., 1996) at the cost of a considerable increase of the complexity of measurements. The magnitude of the underestimation of spinal loading by restricting the analysis to the sagittal plane evidently depends on the asymmetry of a lifting movement. Expressed as a percentage of measured 3-D torques, this underestimation was previously shown to range from 20 % for moderately asymmetric lifting up to 61 % for highly asymmetric lifting (Kingma et al., 1998). The best way to correct for this error is to apply a full 3-dimensional analysis (e.g. (Gagnon & Smyth, 1992; Kingma, Looze, Toussaint, Klijnsma, & Bruijnen, 1996). However, due to the technical difficulty of implementing this solution using currently available techniques, it seems unlikely that full 3-D analysis will become available for large-scale studies in the near future. Therefore it is important to investigate to what extent simple solutions may compensate the underestimation of torques measured by a 2-D model in asymmetrical lifting. It was suggested previously, that a major source of the error in the 2-D analysis is the rotation of the pelvis, resulting in inaccurate projection onto the sagittal plane, of the marker indicating (for instance) the lumbo-sacral (L5S1) joint (Kingma et al., 1998). This problem could be solved by recording a marker at the same location on the other side of the body and by subsequently averaging the two marker positions. A comparable way would be to record the pelvic twist and use this information together with the width of the pelvis to calculate the actual position of the joint in the sagittal plane. Both methods are graphically displayed in figure (1). In the current study it is investigated to what extent correction of the projection error resolves the underestimation of torques by a 2-D model. To this aim, a dynamic 2-D (Looze, Kingma, Bussmann, & Toussaint, 1992b) and 3-D (Kingma et al., 1996) linked segment model were used simultaneously to quantify low back loading during lifting with varying degrees of asymmetry. In addition, information about pelvic twist, obtained from the 3-D model, was used to investigate the effect of correcting the 2-D projection error.

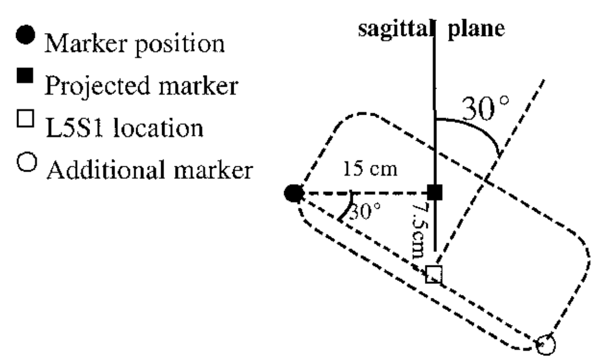

Figure 1. Top view on a pelvis, showing the 2-D projection error of the L5S1 marker for a 30 cm wide pelvis that is twisted 30 degrees. This error can be corrected by averaging marker positions on the left and the right side of the body or by using twist angle and pelvic width information.

METHODS

Subjects, task and procedure

Four male students participated in this experiment after signing an informed consent. In each lift, a 15.7 kg box with handles on both sides was lifted from a shelf 10 mm above the ground, to hip height in upright symmetrical standing position. The initial position of the box was sagitally symmetrical in one condition and rotated 10, 30, 60 and 90 degrees to the right with respect to an axis around both ankles in four other conditions. The initial distance of the box to the right foot was held constant at 5 cm for all conditions. Each lifting movement was performed twice.

Measurements

During the lifting movements, the positions of reflective markers were recorded at 60 samples/s using a 3-D automatic video-based motion recording system (VICON, Oxford metrics). Ground reaction forces were recorded simultaneously by two force-platforms (Kistler, 9218B) and, after analog low-pass filtering at a cut-off frequency of 30 Hz, digitized at 60 samples/s. For both the 2-D and 3-D linked segment model, segment masses and moments of inertia were derived with the aid of anthropometric measurements and regression equations described by McConville et al (1980). Marker positions for the 2-D model and segment center of mass positions for the 3-D model were digitally filtered using a fourth order Butterworth filter with zero phase lag at an effective cut-off frequency of 5 Hz.

2-D model

A 2-D dynamic linked segment model, using inverse dynamics was used to calculate sagittal plane torques at the L5S1 joint (Looze, Bussmann, Kingma, & Toussaint, 1992a). This model used the sagittal plane coordinates of reflective markers attached to landmarks at the fifth metatarsal joint, the lateral malleolus, the lateral femoral epicondyle, the greater trochanter and the L5S1 joint on the left side of the body. Segment angles were calculated as the angle between the line connecting two successive markers and the forward-directed horizontal. The markers represented joint positions. Centers of mass were calculated as a ratio of the distance between two successive markers. Segment linear and angular accelerations were obtained from the time histories of, respectively, the segment centers of mass and the segment angles, by double differentiation using a Lanczos five-point differentiator

3-D model

A seven segment 3-D linked segment model was used to calculate the extending torque at L5S1. This model is part of a full-body LSM, described by Kingma et al(1996). In short: to both feet, lower legs, upper legs and to the pelvis, a cuff constructed of 5 mm thick thermoplastic material (Orfit) was attached. Each cuff contained five spherical markers (10 mm in diameter), mounted with rigid thread. Prior to the experiment, a calibration recording was made for each body segment. For this purpose, additional reflective markers were mounted on the segment at relevant anatomical landmarks in order to allow reconstruction of an anatomical axis system. During calibration, the position of these markers was recorded simultaneously with the markers on the braces. After this recording, the markers on the anatomical landmarks were removed. The markers on the anatomical landmarks were used to reconstruct an anatomical axis system and to calculate the inertia tensor, the center of mass position and joint center position of the segments at the time of the calibration recording. These parameters were transformed for each time instant during the lifting movement using the displacement of the markers on the cuffs (Veldpaus, Woltring, & Dortmans, 1988). In this way, the kinematic input for the 3-D model was generated.
A Lanczos five-point differentiator was used to obtain segment linear accelerations from segment positions and to obtain the first derivative of the inertia tensor. Angular speeds and angular accelerations of the segments were calculated according to Berme et al (1990).
The inverse dynamic process started at both feet, using the data described above and the data from both forceplates, and resulted in time series of the estimated reactive forces and torques at the ankles, knees, hips and L5S1 joint.

Correction of projection error in 2-D model

The error in the estimation of the L5S1 position (due to projection of a marker on the side of the body onto the sagittal plane) was corrected according to figure (1). To this aim, pelvic twist was calculated using the positions of the hip joints, estimated by the 3-D model. A line was drawn through these points and pelvic twist was defined as the angle of this line with respect to the sagittal plane. Using the corrected L5S1 position, the 2-D model was recalculated. In addition, a second correction was applied by dividing the extending torque by the cosine of the pelvic twist angle. In this way, the moment arm of the ground reaction force with respect to the L5S1 joint is partially corrected by taking the twisted orientation of the pelvic flexion-extension axis into account. Taking the total torque, calculated by the 3-D model as a reference value, repeated measures ANOVA's were applied to test for model effects on torque differences. In addition, one-tailed paired t-tests were applied to test for significance of the underestimation of (1) the 3-D extending torque, (2) the uncorrected 2-D extending torque, (3) the 2-D extending torque corrected for L5S1 joint position and (4) 2-D extending toque, corrected for lumboscacral joint position and flexion-extension axis orientation.

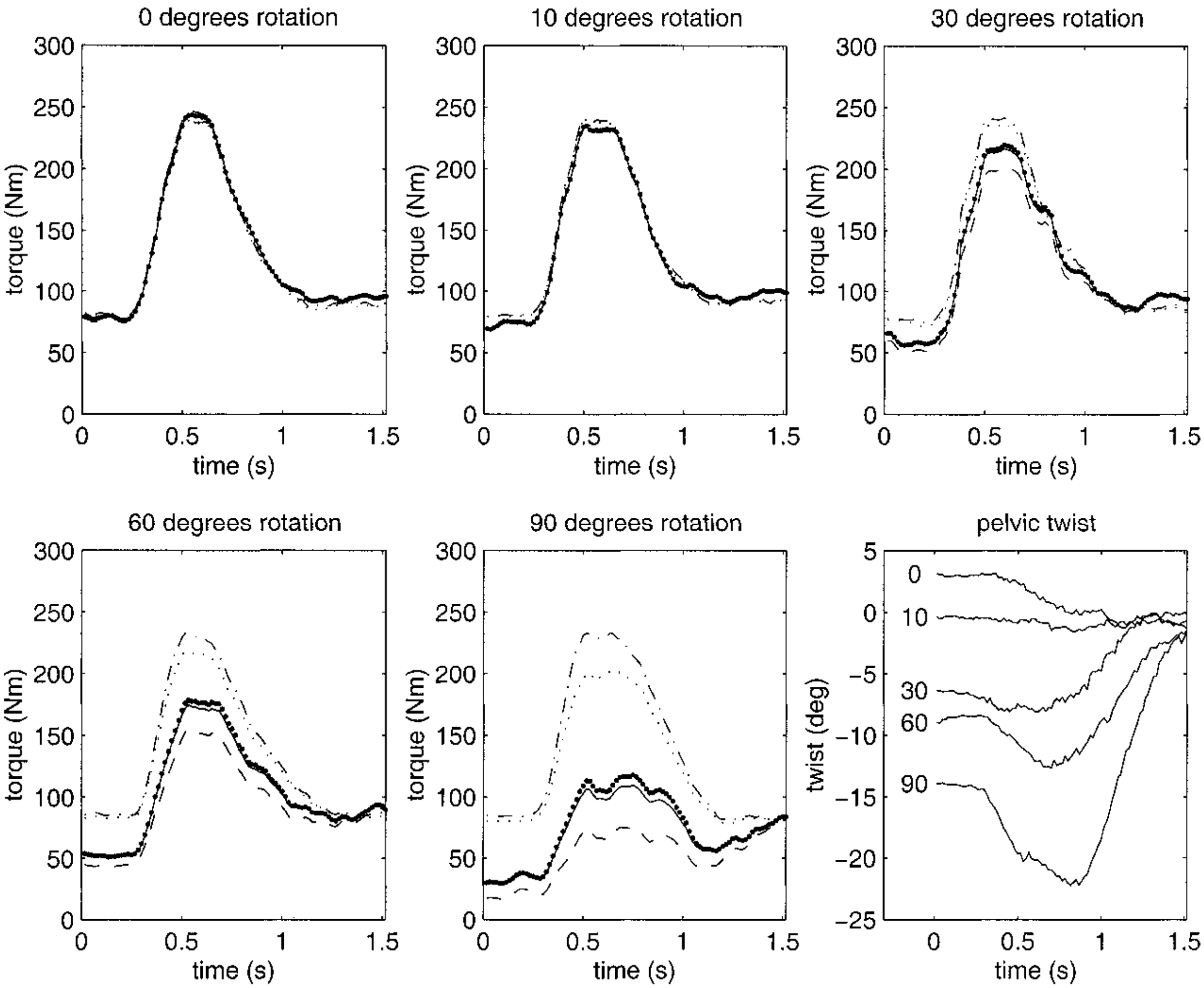

Figure 2: Time series of net torques (averaged over subjects) at the L5S1 joint estimated by the 3-D total torque (dash-dotted), the 3-D extending torque (dotted), the 2-D torque corrected for L5S1 position plus extension axis orientation (dotted, fat), the 2-D torque, corrected for L5S1 position only (solid) and the uncorrected 2-D torque (dashed). The first 5 graphs indicate the different rotation conditions. The last graph (bottom right) indicates the pelvic twist for all 5 rotation conditions.

	3-D total torque		3-D extending underestimation %		2-D extending underestimation %		2-D corrected (1) underestimation %		2-D corrected (2) underestimation %	
Rotation (degrees)	Mean (Nm)	SD (Nm)	mean	SD	mean	SD	mean	SD	mean	SD
0	244.7	55.7	0.4	0.3	-3.3	3.4	-1.7	2.1	-1.9	2.2
10	247.8	48.8	0.6	0.5	2.7	6.3	2.1	3.9	1.7	4.0
30	249.1	46.9	**1.9**	1.5	**16.2**	10.6	**10.3**	7.9	9.1	7.8
60	240.2	41.8	**6.0**	2.6	**32.5**	13.8	**24.3**	10.3	**22.3**	9.6
90	247.9	70.2	**14.4**	6.0	**59.9**	14.9	**50.5**	13.7	**46.9**	13.1

Table 1: 3-D peak total torque and relative underestimation of this torque by the 3-D extending torque, the 2-D extending torque, the 2-D extending torque with correction for L5S1 joint position and by the 2-D extending torque with correction for L5S1 joint position plus flexing-extending axis orientation. Bold numbers indicate significant underestimation ($p<0.05$) according to one-tailed pairwise comparisons.

RESULTS

Repeated measures ANOVA's showed that there was a significant effect of the applied model for all conditions except the 10 degrees rotation condition on the calculated peak torques. Time series of the L5S1 torques (averaged over repeated trials and subjects) showed that differences were only marginal for the 0 and 10 degrees rotation condition (figure 2). Uncorrected peak torques, calculated by the 2-D model, showed a fast increasing underestimation of the total torque when the box to be lifted is rotated out of the sagittal plane, up to 59.9 % underestimation for lifting a box that is rotated 90 degrees (table 1). A comparison of the underestimation percentages between the uncorrected and corrected 2-D extending torques shows that a considerable reduction of the torque underestimation by the 2-D model was obtained by correcting for the L5S1 joint position (Table 1). Still, less than half of the underestimation was corrected in the 30 degrees rotation condition and less than a quarter of the underestimation was corrected in the 90 degrees rotation condition. The additional correction for the orientation of the flexion-extension axis had only small effects (Table 1). This can be attributed to the relatively small amount of twist of the pelvis during rotated lifting (Figure 2, bottom right), causing the cosine of the twist angle to remain close to unity.

DISCUSSION

The current results show that it is possible to keep torque underestimation by a 2-D model, caused by erroneous projection of markers on the sagittal plane, within 10 %, provided that the load that is handled, is not rotated more than 30 degrees with respect to the sagittal plane of the subject. This may be acceptable for many types of research on occupational low back loading. In the current study, information from a 3-D model was used to calculate pelvic twist during lifting. Evidently, this information will normally not be available when 2-D models are used in occupational research. One way to obtain pelvic twist information may be to use a second camera on the other side of the subject. A drawback of this method is that this second camera must be synchronized in space and time with respect to the first camera, resulting in a considerable complication of measurements. A more easy way to obtain information on pelvic twist would be to use a marker on a backward pointing stick on the other side of the body, so that both pelvic markers become visible for the one camera. The horizontal distance between the markers on both sides of the body, together with the known stick length and pelvic width can than be used to calculate pelvic twist.

The remaining underestimation after correcting the 2-D L5S1 torque was still high for the 60 and especially 90 degrees rotation conditions. This is likely attributable to the fact that the point of application of the ground reaction force deviates laterally from the (corrected) pelvic sagittal plane, causing a persisting underestimation of the moment arm of the ground reaction force with respect to the L5S1 joint. This can only be solved by 3-D measurements.

It should not be forgotten that an overestimation instead of an underestimation of torques would be seen when the subject had rotated to the other side, since the projection error than changes its sign. Finally, it should be mentioned that the current study is based on a bottom-up calculated linked segment model. This method depends on the availability of measured ground reaction forces. In a top-down approach, where the linked segment calculations start at the hands, it may be expected that in case of large rotation of the load that is lifted, rotations and their consequent projection errors can be larger compared to the values reported in the current study.

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